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Detection of the fetal ECG during labour by an intrauterine probe

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ABSTRACT

The feasibility of detecting the fetal ECG (FECG) from within the uterus in labour by a technique which is non-invasive to the fetus, has been investigated. The design of a special multi-electrode flexible probe has demonstrated that the FECG can be obtained with signal amplitudes of 20–300 μV and a success rate similar to that of the scalp electrode. Under favourable conditions very large signals can be detected in utero *compared to a scalp electrode, but average signal amplitudes are lower. The probe is suitable as a carrier for other sensors such as pressure and temperature transducers. Currently, simultaneous FECG and intrauterine pressure measurement using a commercially available transducer within the same probe has been achieved.*

Keywords: Fetal and neonatal monitoring, ECG, intrauterine probe

INTRODUCTION

Twenty five years after the introduction of electronic fetal monitoring during labour, there has been little advance in the way fetal heart activity is detected. Most developments have concentrated on improving the electronics and data processing techniques. This has led to the current position where a dedicated and often very sophisticated piece of equipment is capable of producing only two items of data, namely fetal heart rate (FHR) and an indication of uterine activity.

A prospective randomized controlled trial of continuous electronic fetal heart rate monitoring (EFHRM) in labour, involving some 13 000 women, has shown that EFHRM can halve the incidence of neonatal seizures, as compared with a ...d group monitor... stetho-... ...f all neonatal problems, seizure activity ... the closest correlation with long term ...icap[2]. Therefore the value of fetal heart rate monitoring is well established.

...e use of EFHRM has also been widely reported ...ncrease operative intervention rates[3–6]. ...ean section rates have risen sharply in the last ...rs, partly because of an increase in the ...is of 'fetal distress', but also because of a ...ncrease in the diagnosis of 'failure to progress' ...ur. This latter has also been associated with ...M[3,5]. The mechanism for this is unknown ...y relate to increased maternal anxiety.

...present use of separate detectors (e.g. a scalp-... FECG and a catheter — open ended or ...ipped — for intrauterine pressure (IUP), ... separate and sometimes uncomfortable ...inal insertions, is a major disincentive to monitoring during labour. If fetal monitoring is to be extended by the measurement of variables additional to FHR and IUP, the need for the insertion of yet more separate transducers is likely to prove unacceptable to most women.

This paper reports on the feasibility of detecting the fetal heart activity using an intrauterine probe which is non-invasive to the fetus. The probe may also be used as a carrier for other sensors such as pressure and temperature transducers.

THEORETICAL CONSIDERATIONS

Two methods of detecting fetal heart activity from within the uterus are available. First, there is the electrical activity of the fetal heart as it appears either on the fetal skin or in the amniotic fluid near to the fetal skin surface. This may be detected by suitably placed electrodes which are non-invasive to the fetus. Second, the mechanical movement of the heart may be detected by a very sensitive pressure transducer or accelerometer with a good frequency response (a close-proximity acoustic system) or by local Doppler ultrasound methods. The electrical method is preferred for the following reasons.

1. An ECG waveform has a very distinctive character and may be recognized with relative ease in the presence of noise (e.g. EMG).
2. The well defined *R* wave peak of the QRS complex provides an ideal event marker for rate counting.
3. A heart pressure or acoustic pulse would be mixed with other, much larger components induced by fetal and maternal movements including maternal breathing, posture changes, abdominal movements and those due to uterine contractions.
4. The diffuse nature of such a pressure pulse is

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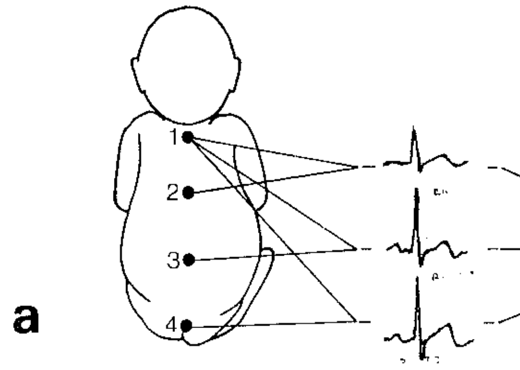
likely to provide no distinctive event marker and 'best fit' correlation techniques would be required to derive a rate. Even then, such techniques would not guarantee accurate determination of the exact time of the event. The same is true for signals derived by Doppler ultrasound.

5. An electrode system is easier to use and more economical than a highly sensitive mechanical sensor or ultrasound system.

A pilot study to examine the feasibility of detecting the electrical fetal heart activity has been carried out[7]. Initially conventional polythene open-ended catheters, as used for IUP measurement, were equipped with stainless steel tips forming an electrode that could be passed into the uterine cavity. A second electrode held within either the vagina or cervical canal served as the reference electrode. In six trials with this type of catheter the FECG was detected on three occasions with amplitudes that ranged between 20 and 300 μV varying not only between insertions but during the period of recording. Signal amplitudes varied from the noise level to some four to six times those normally found on conventional scalp-clip electrodes. Using such a probe it was impossible to position the electrode tip accurately. Electrode to fetus positioning appeared to be critical and may have required actual fetal contact. The maternal ECG (MECG) was also noticed on most of these recordings.

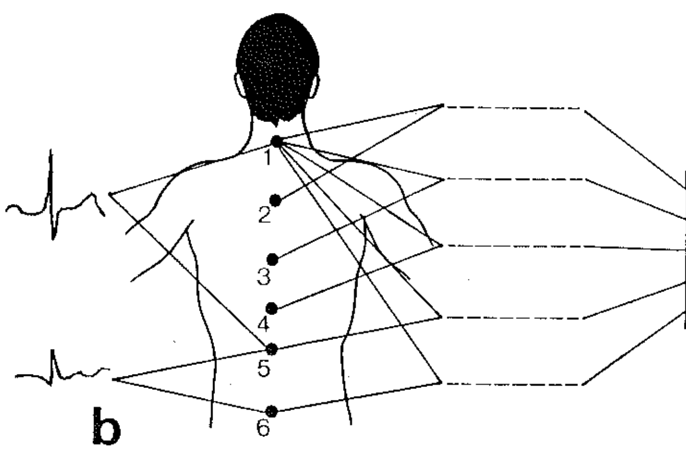
In order to see what signals should be attainable from a fetus we studied the distribution of signal amplitudes from new born babies. As our intention was to insert the intrauterine probe such that it lay along the fetal back we chose a point at the rear of the neck as a reference. Signals were then taken at several points down the spine (*Figure 1a*). This was carried out with the neonate in both wet and dry conditions. In order to mimic the intrauterine environment an attempt was made to obtain ECG signals while the baby was having a salt water bath. In the event this proved too difficult as most babies were too active!

The study demonstrated that a maximum of up to 700 to 1000 μV was available along the dry back of a neonate. The maximum value occurred when the two electrodes were positioned either side of the fetal heart. The amplitude of the signal depended on the electrical axis of the heart and the degree of separation of the electrodes above the surface of the heart. In a similar examination of signal amplitudes along the backs of adults, maximum values of 340 to 800 μV were found. The shape and polarization of the ECG were nominally the same (*Figure 1b*). Signal amplitudes from wet newly-borns, were obtained by lying the baby still covered with amniotic fluid directly on an array of electrodes placed in the base of an incubator cot. The values obtained showed a



a

Electrodes	Spacing (cm)	R wave Amplitude (μV) Dry Skin (n=6) Range	Mean	Wet Skin (n=6) Range	Mean
1–2	6	150–400	240	50–400	213
1–3	12	400–750	550	150–650	350
1–4	18	700–1000	840	250–900	490



b

		Dry Skin (n=7)	
1–2	9	40-75	62
1–3	18	105-350	200
1–4	27	200-300	270
1–5	36	270-425	370
1–6	60	340-800	440

Figure 1 ECG *R* wave amplitudes, with typical complexes, obtained at various points from, **a**, the fetal back, both on dry and wet skin, and, **b**, the male adult back, dry skin only. In all electrode pairs, number 1 is used as the reference. Note polarity of adult ECG between electrodes 5 and 6 (see text)

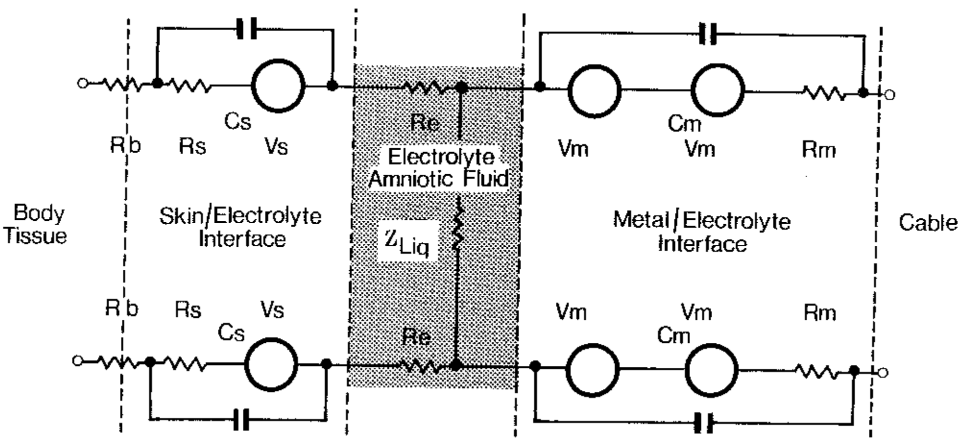


Figure 2 Equivalent circuit for fluid column electrodes. Re, Rs and Rb with Cs is the impedance between electrodes through the sub-dermal tissue. Z_{liq} is the impedance of the electrolyte between the electrodes

variation in maximum values from 250 to 900 μV. The high values were obtained from normal full term vaginal deliveries (one with meconium present) while the lowest signals were from caesarian sections at 36 and 38 weeks where the presence of vernix was [illegible]

The intrauterine environment presents a problem in detecting any electrical activity generated within it. Both the electrical generator (heart) and the detecting electrodes are immersed in an electrically conducting fluid bath (amniotic fluid in the case of the electrodes). There is a conductive path from all parts of the fetal body to the electrodes and between the electrodes themselves. The FECG signal at the electrode surfaces is the result of the vector summation via the amniotic fluid of the potentials over the entire fetal skin and surrounding maternal tissue, and the attenuation produced by the intervening fluid between the electrodes.

A representation of the net effect may be made by modifying an electrical model for two fluid column electrodes[8] by the addition of a lumped inter-electrode impedance (Z_{liq} in *Figure 2*). This represents the bulk impedance of the electrolyte, shorting the potential at the electrode surfaces. In order to maximize the potential at the electrodes the [illegible] of the electrode/skin and inter-electrode (Z_{liq}) impedances must be low.

The impedance from the electrode surface through to the sub-dermal layers of the skin via the fluid column (amniotic fluid) is mainly dominated by the electrode/electrolyte interface and electrolyte/skin impedances which appear in series. In order to minimize this impedance we need to ensure that the electrolyte is in intimate contact with the [illegible] fetal skin. Hair or a waxy vernix layer could have a deleterious effect. Using large area electrodes may help but unfortunately will tend to increase the level of noise induced by mechanical disturbances of the surface charge layer[9].

As the impedance of the electrolyte is relatively low the distance between the electrode surface and fetal skin is unimportant. The presence of a layer of vernix either on the skin or transferred to the electrode surfaces during insertion may be the dominating factor in determining this impedance.

In view of the low impedance of the amniotic fluid, the impedance between the two electrodes can only be minimized by reducing the volume of electrolyte between the two electrodes in an attempt to electrically isolate them. This can be achieved by ensuring that only one face of the electrode is exposed and embedding it in a structure that permits its close proximity to the fetal skin while surrounding it with an area of insulation. In this way if the probe face is squeezed against the fetal skin the electrolyte surrounding the electrode should be expelled, reducing it to a thin film having a relatively high electrical impedance.

In order to observe the above effect, *in vitro* impedances were measured (*Figure 3*). A film was produced by saturating layers of thin tissue paper with electrolyte. As the area and fluid volume was defined the fluid film thickness could be estimated. These tissues were used to connect one electrode to the other, separated by 80 mm. A triangular voltage pulse simulating an ECG *R* wave was fed to the electrodes via a 10 kΩ series resistor. The amplitude of the pulse, across the electrodes, was measured with saline of various concentrations and several samples of amniotic fluid wetting the paper. The results

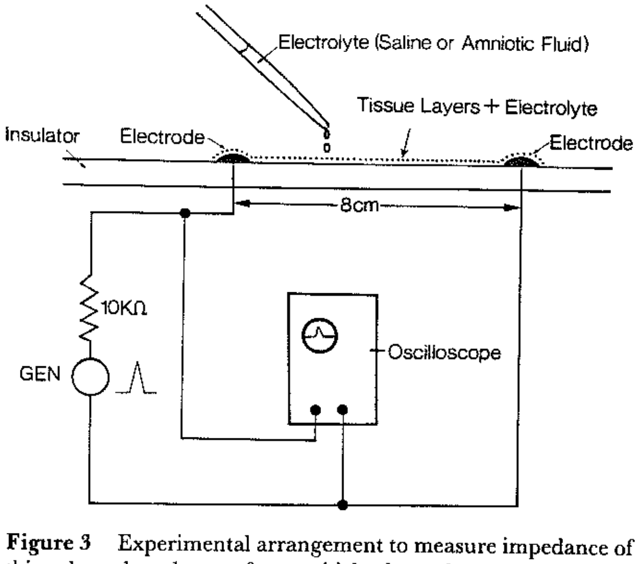


Figure 3 Experimental arrangement to measure impedance of thin electrolyte layers from which the volume resistivity is obtained

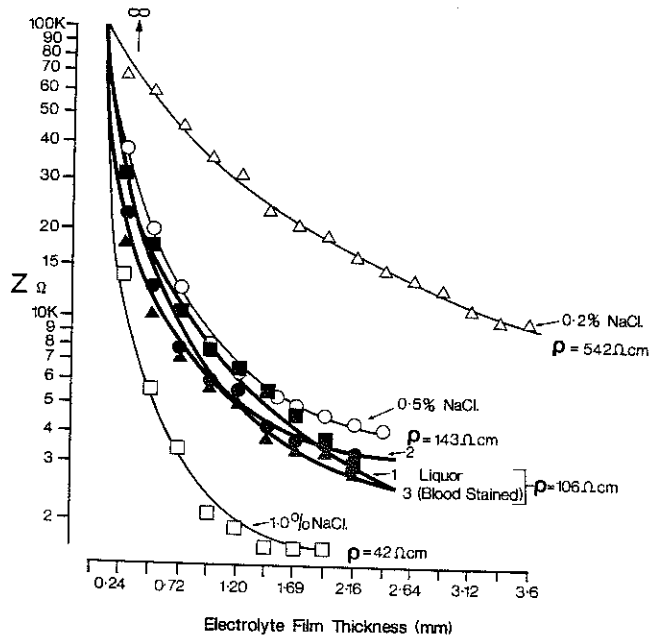


Figure 4 Plot of electrolyte impedance *versus* electrolyte film thickness (1 cm wide) for 18 cm electrode separation. The three amniotic fluid samples are shown with black symbols.

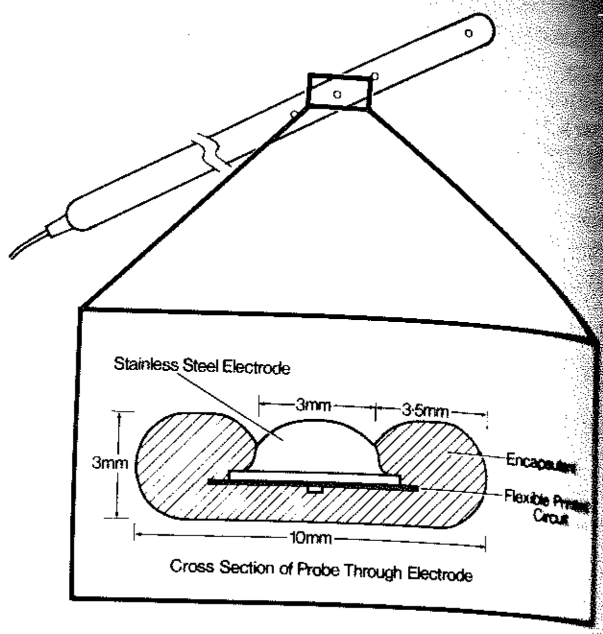


Figure 5 Intrauterine probe with four ECG electrodes. Detail shows cross-section through probe at an electrode.

shown in *Figure 4* indicate that as film thickness increases the impedance between electrodes drops to a level dictated by the bulk impedance of the solution. Amniotic fluid appears, under these conditions, to behave as a 0.6% w/v NaCl solution, suggesting that its electrical impedance is slightly higher than that of physiological (0.9%) saline. As the film is reduced to less than 1 mm in thickness the impedance over the 80 mm length rises rapidly. The mean volume resistivity (ϱ) of the samples of amniotic fluid was calculated as 106 ohm cm. Assuming an electrode to skin impedance of around 1 kΩ, then isolating the electrode from its surroundings with a similar impedance should permit detection of at least 50% of the available signal. For the design outlined below this requires a film thickness of approximately 0.5 mm. Therefore under favourable conditions it should be possible to detect up to 500 μV.

PROBE DESIGN

From both the electrical considerations outlined above and those of the utero-cervical anatomy the design of the intrauterine probe has to take certain criteria into account. The body of the probe must be of a non-conducting and non-toxic flexible material. It must be stiff enough to allow insertion through the cervix and round the fetal head. Measurements taken on new-born babies of neck to buttock distances suggest a probe with an 'active' length of 190 mm. The fetal head, cervix and vaginal canal dictate a total probe length of some 500 mm.

The probe must be shaped so that the orientation of the surface mounted electrodes can be maintained during insertion. A profile with a flat sided oval cross-section will achieve this. The flat sides enable the easy positioning of the electrodes while fulfilling the requirement of surrounding the electrodes with an insulating material. The shape also offers probe flexibility along the surface lying against the fetus while providing a certain amount of transverse rigidity allowing the clinician to have some control over the direction of insertion.

The design of our probe is shown in *Figure 5*. The cross-section is 10 mm wide and 3 mm thick. The overall length is 500 mm. The electrodes are stainless steel domed buttons with an exposed diameter of 3 mm. They are surrounded by insulation which is a minimum of 3.5 mm wide across the probe. These electrodes, numbering three to five, are spread over a distance of 180 mm. Connection of the electrodes is via a strip of flexible printed circuit to which they are mounted. This is embedded into the probe body by an injection moulding technique with a two-part polyurethane compound. The electrodes are slightly raised from the surface of the probe body to provide direct electrode contact with the fetal skin.

The electrical signals on the electrodes of the probe are connected via a lead selector to a small battery-operated low noise amplifier so that any combination of two electrodes can be examined. The amplifier output is coupled via a fibre-optic link to a cassette recording system.

The probe has proved easy to insert even with a cervix dilated as little as 10 mm. In fact it has been noted that insertion at an early stage is preferred while the fetal head is high up. Occasionally difficulties have been experienced once the fetus has descended and engaged in the pelvis. The extra resistance on the probe can cause it to flex and kink. The most favourable position of insertion is such that the probe face lies along the fetal back so that at least one electrode is sandwiched between the fetal head or neck and the cervix. Uterine pressure has

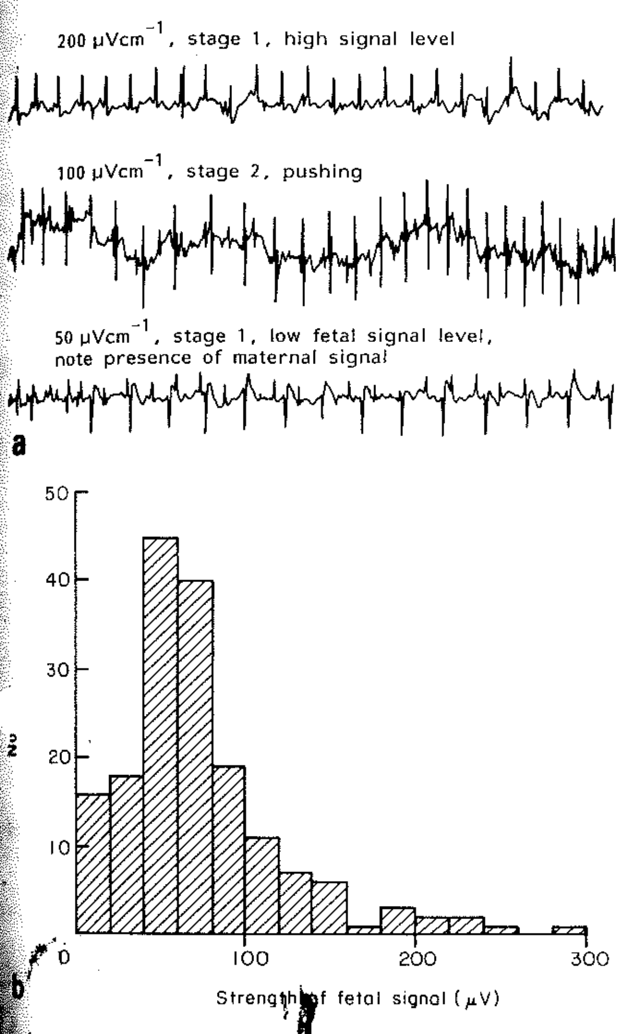


Figure 6 a, shows three traces illustrating various signal levels of both fetal and maternal ECG's obtained from the intrauterine probe. The histogram in **b**, shows the distribution of fetal *R* wave amplitudes found from 14 hours of recordings. While high amplitudes are sometimes detected, most lie between 40 and 60 μV.

been measured satisfactorily with this probe using a conventional sensor tipped pressure catheter embedded into it.

RESULTS

The probe was inserted along a posterior route. A typical signal is shown in *Figure 6a*. Both the fetal and maternal ECGs are usually present. Amplitudes vary widely from case to case and during recording. The distribution of signal amplitudes from 14 h of recordings varies from 20 to 300 μV respectively. The relative amplitudes of the fetal and the maternal ECG complexes also vary so that in any subsequent signal processing to extract the fetal and maternal heart rates, more than simple amplitude discrimination will be required. The success rate (about 90%) of the probe was similar to that of scalp clips. On two occasions the ECG was observed during birth (long after the scalp-clip failed to produce a signal) but the signal amplitude increased markedly during each contraction. This was probably indicative of one or more of the electrodes in use being squeezed onto the fetal head/neck or back thereby increasing the electrical isolation needed for good detection.

No clear correlation was noted with the strength of signal *versus* electrode spacing but overall the electrodes spaced at 120 and 180 mm produced the best result, although the relative polarities of both the fetal and the maternal ECGs varied considerably. One would expect the polarity of the maternal complex to remain constant as the two electrodes should be spaced out on only one side of the maternal heart (see *Figure 1b*–lead position between electrodes 5 and 6). For a head presentation with electrodes spaced out on either side of the fetal heart then maternal and fetal complexes should be of opposite polarity. This was not demonstrated and may be an indication that the probe was not laying in the uterus as imagined.

CONCLUSION

A preliminary investigation into the feasibility of detecting the FECG from within the uterus without piercing the fetal skin has been carried out. The presence of amniotic fluid attenuates the available signals even when intimate electrode to skin contact can be made. Consideration of an electrical model shows that good skin contact and a degree of electrode isolation are required for optimum detection of the high signal levels potentially available. A probe has been designed that allows easy insertion and encourages the conditions for good signal detection. It has been found that both fetal and maternal ECGs can be detected with a success rate of around 90% although signal strengths vary greatly. Further investigations are required to establish the best position and insertion procedure for the probe and to optimize the design of the probe and its electrodes. Special data processing will be required to separate the fetal from the maternal complexes in order to derive both fetal and maternal heart rates. The probe has been used to measure IUP using a commercial transducer embedded within it. Similarly the probe could become a vehicle for taking other sensors into the uterine environment.

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